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published in
XVth Triennial Congress of the International Ergonomics Association
2003

ESTIMATION OF SPINAL FORCES: A COMPARISON OF EMG-BASED AND OPTIMIZATION-BASED MODELS

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Estimates of spinal forces occurring for instance during manual materials handling are derived from biomechanical models. Unfortunately, sensitivity analyses have shown that the model outcomes are quite sensitive to assumptions made, especially regarding antagonistic cocontraction. Optimzation-based models usually predict cocontraction to be absent. EMG-based models take cocontraction into account, but are less practical. The aim of the present study, therefore, was to compare EMG-based and optimization-based estimates of spinal forces in a wide range of work tasks. When comparing mechanical loads across tasks, net moments and estimates of spinal forces obtained using different modeling approaches yield highly consistent results, which suggests that net moments and optimization-based estimates of spinal forces are useful indicators of back load for most ergonomic applications.

INTRODUCTION

Models to estimate spinal forces during for instance manual materials handling rely on assumptions with respect to the distribution of the net moments across muscles spanning the lumbar joints, since these muscles constitute a mechanically undetermined system. Unfortunately, sensitivity analyses have shown that estimates of spinal forces are quite sensitive to such assumptions, especially to assumptions regarding intensity of antagonistic cocontraction (Dieën and Looze 1999).

Two common approaches in estimating spinal compression forces treat the issue of cocontraction quite differently. The first, based on mathematical optimization, estimates the distribution of the net moment by minimizing a function of the muscle forces or muscles activities. Since these cost functions are usually formalizations of some form of efficiency criterion, antagonistic cocontraction is predicted to be absent (Hughes et al. 1995). The second approach uses surface electromyography (EMG) to obtain estimates of forces in individual muscles. The level of cocontraction is in this method estimated from the electrical activity picked up from part of the antagonists. The presence of cocontraction is often used as an argument for using the latter approach (Cholewicki et al. 1995). However, it should be kept in mind that this method is more cumbersome in terms of data acquisition and it still relies on several assumptions, e.g. regarding the activity in deep muscles and regarding the relationship between electrical activity and muscle force. In addition, optimization-based estimations can be implemented in design software, such that estimates of workload can be made before the actual situation has been realized.

The aim of the present study therefore was to compare EMG-based and optimization-based estimation of spinal forces over a range of tasks. Previous studies have made such comparisons for spinal compression forces (Cholewicki *et al.* 1995, Dieën and Kingma 1999, Nussbaum et al. 1999, Dieën et al. 2000), but to our knowledge not for shear forces. In addition, these previous studies dealt with a limited range of tasks only.

METHODS

After signing an informed consent, ten healthy young males (age 24.8 (SD 2.9) yrs, body mass 73.8 (SD 12.8) kg, height 1.80 (SD 0.11) m) participated in a laboratory experiment. Prior to the actual experiment subjects performed three times seven attempted maximum isometric contractions of the trunk muscles as described by McGill (1991). For each muscle, the maximum EMG amplitude over the 21 measurements was considered to be the MVC value. The subjects performed a series of 28 simulated tasks, mainly involving lifting while standing, but also including static pushing and pulling and light seated tasks, such as cash register work.

A dynamic 3-D linked segment model, described in detail by Kingma et al. (1996), was used to calculate net moments at the level of the L5S1 intervertebral disc. The current, slightly modified, model uses anthropometrical data according to McConville et al. (1980), combined with forceplate data and kinematics from markers on cuffs to follow all body segments during movement. To each cuff a 100x100 mm metal plate was attached with a double hinge joint. Four LED markers were fixed to each metal plate. The hinges allowed positioning the metal plates in such a way that optimal visibility of the markers was guaranteed. Marker positions were recorded using Optotrak (Northern Digital, Waterloo ON, Canada) and stored at 50 Hz. For the present analysis data from cuffs placed on the feet, lower legs, upper legs, pelvis and thorax (the latter segment for kinematics only) were used. Different from previous descriptions, we used low pass filtering with a cut off frequency of 10 Hz instead of 5 Hz. In addition, a global equation of motion, as described by (Hof 1992), was used to calculate the moment about L5S1. To the original equation we added the ground reaction moment measured by the force plate. This moment is non-zero around the vertical axis only. The use of this global equation of motion allows the use of one instead of two force plates. Moments were projected onto a pelvis axis system with its origin in L5S1. With the subject standing symmetrically and upright, the x-axis pointed horizontally forward, the y-axis pointed left, and the z-axis pointed vertically upward. The ground reaction force was measured using a large, custom-made, 1x1 meter strain gauge force plate. Force plate data were synchronized to the Optotrak system and stored at 50 Hz.

Twelve pairs of surface EMG electrodes (Blue Sensor; lead-off area 1.0 cm^2, inter-electrode distance 2.5 cm) were attached to the skin after abrasion and cleaning with alcohol. Electrodes were bilaterally attached over the internal oblique, external oblique, rectus abdominis, iliocostalis lumborum (6 cm lateral to L2), and over the longissimus thoracis pars lumborum (3 cm lateral to L1) and pars thoracis (4 cm lateral to T10). Signals were amplified 20 times (Porti-17™, TMS, Enschede, The Netherlands; input impedance > $10^{12}\Omega$, CMRR > 90 dB), band-pass filtered (10-400 Hz) and A-D converted (22bits at 1000 Hz) and stored synchronized to kinematic and forceplate data.

Thorax orientation was described using Euler decomposition of the thorax orientation relative to the pelvis, in the order, flexion/extension, lateral bending, torsion, using the same axis system as for net moments. EMG signals were rectified, low-pass filtered at 2.5 Hz, MVC normalized, and non-linearly transformed into an estimate of muscle activation as described by Potvin et al. (1996). A model of the trunk muscles was used to estimate forces acting on L5S1. All inputs to the trunk muscle model (thorax orientations relative to the pelvis, normalized EMGs, and net moments) were running averaged, reducing the sample frequency from to 10 Hz. The model has been described in more detail previously (Dieën 1997), and consists of anatomical data described by Stokes and Gardner Morse (1995) for the back muscles and by McGill (1996) for the abdominal muscles. After exclusion of the transversus abdominus, lateral part of the external oblique, the psoas major muscle and the latissimus dorsi muscle, the model consisted of 88 muscle slips crossing the L5S1 joint. For muscle slips crossing the L4 and T12 level, nodes were used to keep the distance between those vertebrae and the muscles constant, in order to let the muscles follow the lumbar curvature during motion. Instantaneous orientation of each lumbar vertebra was estimated using interpolation on the basis of the thorax orientation relative to the pelvis.

The moment about L5S1 produced by the 88 muscle slips was estimated as:

$$Mmusc = \sum_{j=1}^{n} \left[ACT_j \bullet \sigma_{\max} \bullet pcsa_j \bullet \omega_j \bullet \delta_j \right] \times ri_j$$

with ACT_j being muscle activation either estimated from the normalized EMG signal allotted to muscle slip j, or derived by static optimization, σ_{max} being maximum muscle stress, $pcsa_j$ the physiological cross-sectional area of muscle slip j and ω_j and δ_j correction factors for the contraction velocity and instantaneous muscle length of muscle slip j, and finally ri_j being the vector from muscle insertion to the center of the L5S1 disc. The correction factors ω_j and δ_j are based on dynamical properties of human and animal muscles as described by van Zandwijk (1998) and passive length tension properties as described by Woittiez et al. (1984). The muscle lengths and contraction velocities were calculated on the basis of three dimensional trunk angles. In the EMG driven model (EMGMOD), maximum muscle stress was estimated iteratively using a least squares fit between the 3D muscle moment and the 3D net moment derived from the linked segment model. In the optimization-based model, maximum muscle stress at optimum length was set at 61 Ncm^{-2}, which was the average value found with the EMG driven model. Muscle activation was estimated as described in (Dieën 1997). In short, upper and lower bounds on activation were 0 and 1, equality between 3D muscle moments and net moments functioned as a constraint. Activation levels of muscles were constrained to be constant among the muscles driven by one of the 12 EMG signals in EMGMOD. The cost function was ΣACT^3_j. This cost function not only preferentially recruits muscles with a large moment arm, but also favors muscles with a large cross-sectional area. This model will be referred to as OPTA3MOD. In addition we performed the optimization using as cost function: $\Sigma ACT_j \bullet pcsa_j)^3$. This function penalizes the activation of larger muscles. It is roughly equivalent to using a cost function based on muscle force to a power of 3, but since it does not incorporate the effects of muscle length and shortening velocity on force, it is more closely related to active muscle mass. This model will be indicated as OPTF3MOD. All optimization was performed using the Matlab optimization toolbox (The Mathworks Inc., Nattick, USA).

The data analysis focuses on a comparison of force estimates from the EMG driven model (EMGMOD) and the optimization models (OPTA3MOD and OPTF3MOD). Force estimates were summed over muscles and the maximum of the vectorial sum of the resulting force vector was determined in each task in each subject as were the maximum values of the compression and anterior shear components with respect to L5S1. We only considered forces produced by the muscles, since the inertial components were estimated using inverse dynamics and hence were equal for the three models. Finally, we analyzed the relationship between the maximum value of the vectorial sum of the 3D net moments and the spinal force estimates.

RESULTS

In general, the EMG model yielded a good fit of muscle moments to the net moments. The median coefficient of determination (R^2) of the fit of muscle moment to net moment was 0.87 over all three dimensions. This coincided with a median absolute difference between muscle moments and net moments of 11 Nm. Maximum tension estimates averaged within subjects ranged from 29 to 108 Ncm^{-2}, with a mean value of 61 Ncm^{-2}. Gain estimates were reasonably consistent within subjects.

For most ergonomics applications of the models discussed here, results at the individual level are not important. We therefore present results of group averaged data only. Total muscle forces from both optimization models were highly correlated to those from EMGMOD. Coefficients of determination were 0.974 for OPTA3MOD and 0.956 for OPTF3MOD. In addition, the average differences were only 49 (SD 268) N and -281 (445) N.

After decomposing the total muscle force into compression and shear components we again found strong correlations between outcomes from EMGMOD and both optimization models. For OPTA3MOD coefficients of determination were 0.976 and 0.564 for compression and shear, respectively, with average differences being 28 (SD 259) and 560 (SD 187) N. For OPTF3MOD the corresponding values were 0.956, 0.975, -297 (SD 449) N, and 73 (SD 45) N, respectively.

Group averaged peak values of total force, compression force, and shear force estimates were closely correlated to the peak value of the total net moment. For EMGMOD coefficients of determination were 0.979, 0.979, and 0.811 respectively. For OPTA3MOD the corresponding values were 0.996, 0.996 and 0.344 and for OPTF3MOD 0.976, 0.974 and 0.813. Remarkably the peak force estimates for the 4 asymmetric lifting tasks studied did not deviate from the regression line determined over all 28 tasks.

DISCUSSION

We found only small differences between EMG-based (EMGMOD) and optimization-based (OPTA3MOD) estimates of total and compression forces in a range of tasks, which was in line with our previous studies (Dieën and Kingma 1999, Dieën *et al.* 2000) using a similar model. In contrast, other authors (Cholewicki *et al.* 1995, Nussbaum *et al.* 1999) have reported more substantial differences in compression estimates between optimization and EMG-based methods. Since the model geometry in one of these studies appears fairly similar, the explanation for this disparity probably is the difference in cost functions used in the optimization. In these previous studies, the maximum muscle stress and the compression force were subsequently minimized (Cholewicki *et al.* 1995, Nussbaum *et al.* 1999). This cost function will in some tasks predict substantially lower compression forces than Σact^3, which was used in the present study (Hughes 1995, Dieën and Kingma 1999). The latter was shown to make more valid predictions of muscle activity (Hughes et al. 1994, Dieën 1998, Dieën and Kingma 1999). In addition, we found OPTF3MOD to provide similar predictions of shear forces as EMGMOD. The cost function in OPTA3MOD not only preferentially recruits muscles with a large moment arm, but also favors muscles with a large cross-sectional area. Consequently, the lateral part of the internal oblique was predicted to be much more active in OPTA3MOD than estimated from the EMG amplitude. This muscle has a small extension moment arm, but large cross-sectional area in our model. The preferential recruitment of the lateral internal oblique strongly affected the direction of the muscle force vector and thereby the shear component. Due to geometric considerations, the compression component was much less affected.

A strong relationship was found between the peak 3D net moments and peak spinal force estimates. This suggests that a single-equivalent model (SEM) could be used to analyze these tasks. The regression coefficients of the relationships between net moment and compression estimate indicate that an effective moment arm of about 0.05 for this SEM would be appropriate. This is slightly higher than previous suggestions based on the same model geometry (Dieën and Looze 1999). The difference probably is an effect of improved representations of the curved lines of action of the muscles in the present version of our model. The estimate moreover was comparable to that based on imaging studies (Reid and Costigan 1987, Chaffin et al. 1990). In comparison to other studies (McGill and Norman 1987, Potvin et al. 1992) this effective moment arm is however rather small. Differences between tasks (degree of trunk flexion) may have contributed to this disparity, but in addition our model appears rather small (Dieën and Looze 1999). In contrast to our expectations (Dieën and Kingma 2001), the relationship between net moments and spinal forces held up in asymmetric lifting suggesting that the vectorial sum of the 3D net moment in a pelvis axis system is a valid indicator of spine load in asymmetric lifting. Praagman et al. (2000) similarly reported a close relationship between shoulder joint compression and net moments.

In conclusion, when comparing mechanical loads across tasks, net moments and estimates of spinal forces obtained using different modeling approaches yield highly consistent results. These findings suggest that net moments and optimization-based estimates of spinal forces are useful indicators of back load for ergonomic applications.

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